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L17: Entry 11 of 17 File: USPT Apr 27, 1999

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TITLE: Single circuit ladder resonator quadrature surface RF coil

Abstract Text (1):
A single-circuit quadrature surface coil is formed from two ladder resonator coils and includes a first mode circuit path for detecting or generating magnetic flux in a vertical axis from a body under investigation and a second mode circuit path for detecting or generating magnetic flux in a parallel axis, with the first mode and second mode currents 90 degrees out of phase. The surface coil, which supports two resonance current modes for quadrature operation on only one single coil conductor structure, provides a high signal-to-noise ratio (SNR) and a good B.sub.1 homogeneity over the imaging volume. This coil alone may be used either for both transmitting and receiving RF signals or for detecting RF signals as "receive only." This coil is well suited for imaging the human neck, spine and heart.

Brief Summary Text (2):
present invention pertains generally to Magnetic Resonance imaging (MRI) apparatus, and more particularly to a quadrature surface coil for use with MRI apparatus.

Brief Summary Text (4):
MRI provides a unique non-invasive imaging method for discriminating the main components of human disease pathology. As a result, MRI is one of the most widely used diagnostic imaging tools in today's hospitals throughout the world. A typical MRI system includes a main magnet to generate a uniform DC magnetic field, three gradient coils to generate linear and orthogonal magnetic field gradients, a transmitting and receiving radio frequency (RF) antenna to generate imaging pulses and receive the resulting RF emissions, and an operator interface and control station. For human imaging the magnet is mainly superconducting in nature and has a cylindrical shape, although at the present time open "C" arm magnet geometries are also used for imaging the human body. For higher strength magnetic fields (0.5 T and higher), the superconducting magnet is used to generate a highly uniform static magnetic field with a clear bore diameter of 90 cm or larger for human patient access.

Brief Summary Text (5):
Gradient coils are electromagnetic coils capable of generating linearly varying and axially directed static magnetic fields along the three spatial directions (x,y,z) of a Cartesian coordinate system. The function of each one of the three orthogonal gradients is to encode the spatial information as a frequency or phase variation. In general, higher gradient strengths of 40-100 mT/m with faster rise times of 40-150 .mu.sec at full strength are required for faster imaging techniques. Standard methods for production of linear magnetic field gradients in MRI systems consist of driving discrete coils with a current source of limited voltage. The discrete coils are wound in a bunched or distributed fashion on an electrically insulating hollow light cylinder coil-form.

Brief Summary Text (6):
A digital radio frequency transmitter transmits radio frequency pulses or pulse packets to a whole body RF coil to deliver RF pulse into the examination region. The RF pulses are used to excite, prepare, saturate, invert, refocus, or manipulate the resultant bulk magnetization due to ensemble average of the magnetic moment of a

specific nuclear spin such as proton in selected portions of the examination region. For whole body applications, the resonance signals are commonly picked up by the same whole body RF coil. For other more regional focused applications, the signals are often picked up by local coils placed at the vicinity of the examination region other than the whole body coil. Alternatively, a receive-only coil can be used to receive resonance signals induced by the body RF coil. For example, in human head imaging, an insertable head coil can be inserted surrounding a human brain at the isocenter of the magnet used for receiving the RF signal. Conventional RF coils for MRI application are the birdcage coil, single loop surface coil and surface array coil.

Brief Summary Text (7):
In MRI, the resultant radio-frequency signals, which are spatially encoded, are picked-up by the receiver RF coil, amplified and then demodulated/digitized by a receiver. A sequence controller controls or schedules the timing sequence of the three orthogonal gradients, RF pulse waveforms, frequency offset, RF phase, data sampling window of the receiver, as well as other events such as triggering to generate a variety of MRI sequences, such as spin echo imaging, gradient echo imaging, fast spin echo imaging, and echo planar imaging. An image reconstruction processor sorts the spatially encoded image data according to the order in which they are received and transforms the data to form the final MR image.

Brief Summary Text (8):
More specifically for RF antenna, a simple conductor loop interrupted with some capacitors of proper values can serve as a local RF antenna that is capable of transmitting RF signal to its vicinity and detecting minute RF signal from its vicinity with a relatively high signal-to-noise ratio (SNR). Such a coil can only transmit and detect one component of the magnetization signal during M imaging, and often is referred as a linear (RF) coil. Since the detectable magnetization for MRI is a two dimensional vector, to improve the SNR and B.sub.1 uniformity, a quadrature version which receives two orthogonal components simultaneously was then introduced for the same geometry configuration.

Brief Summary Text (9):
PRIOR ART QUADRATURE RF COILS

Brief Summary Text (10):
Previously, there were some early quadrature surface RF coil designs described by various people including Hyde, et al., U.S. Pat. No. 4,721,913, issued Jan. 26, 1988, and Mehdizadeh, et al., MRM, 1,256,1988, U.S. Pat. No. 4,918,388, issued Apr. 17, 1990. Both the Hyde and Mehdizadeh designs consist of two separated and isolated coil circuits which simultaneously generate or detect two orthogonal RF polarizations independently. Furthermore, it has been shown that the two circuits can be made mutually decoupled from each other for the respective resonance modes of interests by means of proper geometrical overlapping. The two signals obtained from two orthogonal polarizations can be combined to enhance the SNR in the resulting image using a quadrature combiner. Boskamp et al., in U.S. Pat. No. 5,030,915, issued Jul. 9, 1991, disclose a single circuit quadrature RF coil formed from a pair of parallel mode single loop coils. The Boskamp et al. coil is formed from an annular outer conductor with an inner conductor connected on first and second ends at respective spaced points around the outer conductor.

Brief Summary Text (11):
Another approach to increase both sensitivity and uniformity for a surface coil over a certain coverage was the "half birdcage" design proposed by Ballon et al., JMR, 90, 131-140 (1990). The half birdcage design is also known as a one dimensional ladder resonator. Due to the multiple conductor design of the half birdcage design, it provides better uniformity of field than single loop coils. Although it is an improvement over the single loop design, it is not a quadrature design, lacking the superior SNR of the quadrature RF coils.

Brief Summary Text (14):
The present invention provides a ladder resonator quadrature coil having the superior SNR of quadrature coils and the uniformity of ladder resonator coils. According to one embodiment, the coil of the present invention comprises a

single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path. According to one aspect of the invention, the coil construction provides a first mode circuit path sensitive to magnetic flux in a first orientation, and a second mode circuit path sensitive to magnetic flux in a second orientation orthogonal to the first whereby quadrature operation is obtained. According to another aspect, first and second relatively isolated signals representing orthogonal magnetic fields emitted from a body under investigation in an MRI apparatus are obtainable from the respective first and second mode circuit paths. This embodiment, which supports two resonance current modes for quadrature operation on only one single-coil conductor structure, provides a high signal-to-noise ratio (SNR) and a good B.sub.1 homogeneity over the imaging volume. This coil embodiment may be used either for both transmitting and receiving RF signals or for detecting RF signals as "receive only". This embodiment of the invention is also well suited for imaging the human neck, spine and heart.

Brief Summary Text (15):
According to another exemplary embodiment of the invention, the single circuit ladder resonator quadrature coils of the present invention are mounted in combination with a standard birdcage coil to provide additional imaging capability for the neck region of a user having their head positioned in the birdcage coil. According to yet another embodiment, the coils of the present invention are integrated or mounted in combination with other conventional coils to provide supplemental imaging capability.

Drawing Description Text (5):
FIGS. 3 and 4 illustrate simplified schematic circuit diagrams of embodiments of the ladder resonator quadrature surface coils of the present invention with circuit elements for decoupling the coils when used in a receive only mode.

Drawing Description Text (6):
FIG. 5 illustrates a pair of ladder resonator quadrature surface coils of the present invention mounted in combination with a birdcage coil.

Drawing Description Text (7):
FIG. 6 illustrates the coils of FIG. 5 deployed in an MRI apparatus.

Detailed Description Text (2):
In the following detailed description of the preferred embodiments, references made to the accompanying drawings which form a part hereof, and in which is shown by way of illustration specific preferred embodiments in which the invention may be practiced. These embodiments are described in sufficient detail to enable those skilled in the art to practice the invention, and it is to be understood that other embodiments may be utilized and that structural, logical, physical, architectural, and electrical changes may be made without departing from the spirit and scope of the present invention. The following detailed description is, therefore, not to be taken in a limiting sense, and the scope of the present invention is defined only by the appended claims and their equivalents.

Detailed Description Text (3):
The ladder resonator quadrature coil of the present invention combines two separate ladder resonator coils for two polarizations into an integrated coil conductor structure. This new composite coil provides both high SNR and good B.sub.1 homogeneity over the imaging volume. The simplified topological structure of an embodiment of the new ladder resonator quadrature coil design of the present invention is shown schematically in FIG. 1. For simplicity, the decoupling circuits for receive only operation, which are required by the standard MRI system to decouple the coil from the whole body coil by detuning the resonance frequency of the coil during RF transmission, are omitted from the illustration of FIG. 1. As shown in the FIG. 1, the ladder resonator quadrature coil embodiment shown therein can be said to be formed of two ladder resonator coils sharing a common conductor pattern which is symmetrical about a center conductor. In the embodiment of FIG. 1, two parallel conductor strips in the horizontal direction are connected by five (5) or more uniformly spaced vertical oriented conductor strips or legs, also referred to herein as "paths." Both horizontal and vertical conductor strips are interrupted

by capacitors. The coil structure is symmetric with respect to the center vertical conductor, or circuit path, to provide an annular first mode circuit path using the two (2) or more spaced conductor legs on each side of the center conductor leg. The circulation of current through this path is shown in FIG. 2A, wherein the dotted center line corresponds to the center leg conductor of coil 1. The first mode circuit path detects or generates magnetic flux of a first polarization from a body under investigation. A second mode circuit path including the center circuit path, the current flows for which are shown in FIG. 2B, detects or generates magnetic flux in another plane of polarization, with the currents in the first and second mode circuit paths 90 degrees out of phase.

Detailed Description Text (4):
Electrically the overall coil embodiment 1 is nothing but a simple one dimensional circuit network for current loops. Such a structure supports many resonance current modes (there are four independent modes), whose mode currents share the same conductor pattern. The current patterns shown in FIG. 2A and 2B for these two useful resonance modes of coil 1 are capable of generating fields either vertical (perpendicular) or horizontal (parallel) to the surface of the coil. Although the frequencies corresponding to the two specific modes are often different, in the coils of the present invention these two frequencies are made identical by introducing a set of capacitors cross both center vertical conductor (DC) and one of the two end-ring conductors (AC, CB). In order to accomplish the required frequency matching between two polarizations for quadrature operation, the capacitance of the capacitor on the center vertical conductor is varied or set accordingly. In addition, the values for different capacitor components can be predicted for a given structure using a numerical model for the RF coil. At the resonance frequency, two independent RF signals corresponding to the two orthogonal components of the excited magnetization vector can be taken out simultaneously directly from the coil for the purpose of MRI. One RF signal can be taken out from the center vertical conductor strip or leg through a direct electrical coupling cross the capacitor, and the other can be taken out at the center of one of the parallel conductor strips in similar fashion. For example, U.S. Pat. No. 5,030,915 illustrates in more detail how these signals can be taken off of a single circuit quadrature coil. Furthermore, these two useful resonant modes are intrinsically isolated from each other as a result of their corresponding geometric shapes. Other means of tuning the coils are also possible such as also shown in U.S. Pat. No. 5,030,915, and in particular FIGS. 4, 5 and 6 of that patent.

Detailed Description Text (5):
According to one embodiment of the invention, the conductor pattern of the coil 1 can be constructed on an electrically insulated as well as heat resistant former using copper strips etched or applied to the flexible former. The size of the coil can be made larger or smaller depending on the required coverage of a specific application. The face of the coil can be made to be flat, curved or flexible. Using a construction of this type, approximately-21 db isolation (loaded) between the two polarizations is achievable. The two signals corresponding to two orthogonal polarizations were transmitted independently from the coil through two short flexible 50 ohm coaxial cables to corresponding low noise pre-amplification modules. Then, the amplified signals were fed to a multiple channel port on an MRI apparatus for transmission, demodulation, digitization and image reconstruction. With the decoupling circuits as shown for example in FIG. 3 incorporated in a coil of this construction, the resulting coil in a receive only mode was used for imaging studies at 1.5T on a Siemens VisionTM whole body MRI system. While the coil 1 is illustrated in FIG. 1 as having a rectangular shape, it can be any arbitrary shape that preserves the symmetry necessary for tuned operation, such as the general shapes shown in FIGS. 7a, 7b and 7c of U.S. Pat. No. 5,030,915.

Detailed Description Text (6):
With a proper impedance match circuit the two RF signals obtained from the coil structure of FIGS. 1-4 can be pre-amplified and combined using a standard analog quadrature combiner. Alternatively, for achieving the optimal SNR, the MR signals corresponding to two orthogonal polarizations from the coils can each be separately pre-amplified using a low noise RF amplifier and fed into two RF receiver channels, so the two signals are demodulated and digitized independently. The required quadrature signal combination can be accomplished digitally in the image domain

using a software combination accompanied with phase correction for compensating the coil reception phase variation. In addition, the shape and geometrical size of the coils made in accordance with the invention can be optimized for a given application.

Detailed Description Text (7):
Furthermore, ladder resonator quadrature coils made according to the present invention can be used completely alone for both transmitting and receiving RF signal as a transmit -receive (T/R) coil for MR imaging and spectroscopy. Using this design, the traditional half -birdcage coil can be modified to be operated in quadrature mode. One of the advantages of the T/R coil design is its relatively simple coil structure, since it does not require decoupling components on the coil; another advantage is the 40% improvement in SNR over conventional birdcage coils.

Detailed Description Text (8):
If a ladder resonator coil made in accordance with the invention is used in a receive-only operational mode, the coil needs to be made invisible to the transmit RF coil during RF power transmit by means of detuning the resonance frequency of the receiver coil away from that of the transmitter coil, which may be a saddle coil or full birdcage coil by way of example. There are various possible approaches suitable for this detuning. Using a few PIN diodes and inductors in proper locations on the coil 1 structure, the resonant frequencies for the two useful modes of coil can be detuned by selectively switching these diodes on or off using an externally applied voltage signal across these diodes. Examples of coils 1' and 1" modified for this purpose are shown in FIGS. 3 and 4. In FIG. 4, tank circuits activated with the PIN diodes are used to detune the circuit, while in the embodiment of FIG. 3, PIN diodes alone are employed. Of course, many other configurations of circuit elements and position can be used to detune the coil, and the examples given herein are not to be construed as limiting.

Detailed Description Text (9):
For many practical applications, ladder resonator quadrature surface coils 1 (or 1' or 1") can be integrated into or mounted in combination with many other conventional coils available on clinical MRI systems, one of which is a standard birdcage head coil 2 as shown in simplified form in FIG. 5. Such an "integrated" coil set is highly desirable for head-neck imaging applications. This allows the possibility of performing a complete head-neck examination without changing the RF coil in the middle of examination. The different elements can be selectively switched on and off (i.e. detuned) by the means of a DC voltage signal applied to the coils externally. In FIG. 5 coils 1 (or 1' or 1") are shown mounted or integrated with the full birdcage coil 2. Mounting or integration can be accomplished by mechanically or otherwise fixing the location of the coils 1 in relation to the birdcage coil 2 so that, in this case, when a patient's head is positioned in the birdcage coil, their neck is positioned in the sensitive region of the coils 1. By constructing the ladder resonator coils with input and output leads compatible with existing MRI systems, these coils can be made to work with such existing systems, and mounted to work in conjunction with conventional full birdcage coils of various types or manufacture, or in conjunction with other imaging coils commonly used with MRI systems.

Detailed Description Text (10):
FIG. 6 illustrates the use of the coils of the present invention as shown in FIG. 5 when deployed in an MRI apparatus. The MRI apparatus shown in FIG. 6 includes a plurality of superconducting main magnetic field coils 10 to generate a generally uniform static magnetic field along a longitudinal or z-axis of a central bore 12. The superconducting coils are mounted on a dielectric former 14 and received in an annular, helium vessel 16. The helium vessel is filled with liquid helium to maintain the superconducting magnets at their superconducting temperature. A main magnetic field shield coil assembly 18 generates a magnetic field which opposes the fields generated by the main magnets 10 in regions surrounding the superconducting magnets 10. The annular helium reservoir 16 is surrounded by a first cold shield 20 which is maintained at about 20.degree. K. or less. A second cold shield assembly 22 is chilled to about 60.degree.-70 .degree. K. or less. An outer vacuum vessel 24 encases the cold shields to define a vacuum reservoir therearound. Layers of mylar insulation 26 are arranged between the vacuum vessel 24 and the cold shield 22.

Detailed Description Text (11):
A circularly cylindrical, whole body gradient coil assembly 30 is mounted on a circularly cylindrical dielectric former and mounted within the bore 12. A circularly cylindrical, whole body RF coil 32 is mounted on a circularly cylindrical dielectric former and mounted within the bore 12. A circularly cylindrical dielectric cosmetic sleeve 34 shields the RF and gradient coils from view and protects them from damage. The ladder resonator quadrature coil/birdcage head coil combination shown in FIG. 5 is positioned in bore 12 in close proximity to the patient. As illustrated, the ladder resonator quadrature coils 1 (or 1' or 1") and the birdcage coil 36 (in this case a quadrature birdcage coil) each include separate circuits connecting them to the transmitter and receivers. Only one ladder resonator quadrature coil is shown in FIG. 6, but preferably a second is provided on the other side of the patient's neck as shown in FIG. 5, and an additional circuit is provide to transmit and receive RF from the other coil. Alternatively, a switch could be provided to switch between the leads from the two ladder resonator quadrature coils so that only one set of transmit and receive circuits would be needed.

Detailed Description Text (12):
A transmitter 40 is connected with the whole body RF coil 32 for transmitting resonance excitation and manipulation pulses thereto. Preferably, a quadrature divider 42 splits the radio frequency signal into two components and phase shifts one component 90.degree. relative to the other. The two components are applied in quadrature to the whole body RF coil. A gradient control means 44 is connected with the gradient magnetic field coils 30 for providing current pulses thereto for generating magnetic gradient pulses across the examination region. A sequence control means controls the radio frequency transmitter 40 and the gradient control means 44 to generate conventional resonance excitation sequences such as spin echo, gradient echo, field echo, sequences and the like. Preferably, resonance is excited in two preselected planes or slabs by applying a linear z-gradient field concurrently with a tailored radio frequency excitation pulse for exciting resonance in the two or more preselected slices or slabs. Preferably, one of the slices or slabs intersects the region examined by each of the surface coils 1 (or 1' or 1") and 38.

Detailed Description Text (13):
The radio frequency transmitter 40 and the gradient control 44 under the control of the sequence control 46 elicit simultaneous magnetic resonance responses in planes or slabs through each of the quadrature surface coils 1 (or 1' or 1") and 38. The signals from the two quadrature surface coils are conveyed to a pair of quadrature combiners 50, 52. The quadrature combiners impose a 90.degree. phase shift on one of the detected quadrature components and combine the components. Preamplifiers 54, 56 amplify the signals before they are received by a receiver means 60, such as a pair of digital quadrature receivers 60.sub.1, 60.sub.2, which receive and demodulate the resonance signals. An interface circuit 62 includes analog-to-digital converters 64, 66 for digitizing each received resonance signal to generate a digital data line.

Detailed Description Text (14):
The digital data lines are stored in data memories 70, 72 of a computer means 74. An image reconstruction means such as an inverse two-dimensional Fourier transform means 76 reconstructs sets of data lines from the data memories 70, 72 into electronic digital image representations which are stored in image memories 80, 82. A video processor means 84 converts the digital image representations into the appropriate video format for display on a video monitor 86 or other human-readable display.

Detailed Description Text (15):
Those of skill in the art will recognize that the conductors of the ladder network of the birdcage coil 38 and quadrature coil 1 must be oriented in parallel with the uniform magnetic field, such that use of the coil system with different orientations of the uniform, main field requires reorientation of the coils to meet this requirement.

Detailed Description Text (16):
When the coils of the present invention are used for receiving in the embodiment or

arrangement shown in FIGS. 3 and 4, the coil advantageously improves SNR by more closely allowing the correlation of the geometric location of the transmit pulses with the corresponding RF emissions detected by the coils . Furthermore, according to another embodiment, a sensor is provided to detect the pulse in the carotid artery, so that the imaging pulses can be synchronized or triggered on pulse events, and thereby provide repeatable, consistent imaging.

Detailed Description Text (17):
As suggested above, ladder resonator quadrature coils constructed according to the present invention can be of a substantially planar configuration in the manner of a surface coil, or shaped or curved to provide a half -birdcage coil configuration, providing a single half -birdcage coil structure operable in a quadrature mode. In terms of clinical imaging application, coils constructed according to the present invention are well suited for imaging the human neck, spine and heart. For multi-nuclear spectroscopy applications, such coils can be easily made double tuned for simultaneously observing two frequencies.

Detailed Description Text (18):
In the invention disclosure, for simplicity and the purpose of description, the ladder resonator quadrature coil embodiments are shown with five legs, but the number of legs is arbitrary, and can be more than five. However, the term "ladder resonator quadrature coil " as used herein requires that each respective coil on either side of the circuit include multiple legs or "vertical" conductors, such that each respective coil itself forms a ladder. This structure distinguishes the design of the present invention from that of U.S. Pat. No. 5,030,915, wherein the respective sides of the single circuit quadrature coil are single loop coils. Thus, while it is required that the ladder resonator quadrature coil of the present invention include multiple legs on each side, the number of legs as noted may be greater, such as 7, 9, 11 etc. Alternatively, the number of legs may be even. In the case of even numbers of leg conductors, the center circuit path is formed from two (or more) conductors used as an effective center leg. Furthermore, the spacings between vertical legs or conductor strips can be varied symmetrically with respect to the center strip.

Detailed Description Text (19):
Further possible modifications to the invention include placing the capacitors on the other end-ring to preserve symmetry. Also, as is well known, inductive coupling can be used for feeding and taking signals out from the coil, or the orthogonal signals can be combined using a standard quadrature hybrid combiner before or after pre-amplification. It is also possible that the two orthogonal modes of the coil can be intentionally tuned to two different frequencies, providing a double tuned coil for multi-frequency or multi-nuclear imaging or spectroscopy.

Detailed Description Text (20):
In addition to the above noted variations in coil design, coils constructed according to the present invention can be made to conform to the geometry of the anatomy using curved (non-straight) vertical and horizontal conductors, or be made to have more open space for the interventional purpose as well as conventional imaging. Furthermore, two or more small coils can be integrated into a ladder resonator quadrature coil of the present invention for active position and orientation tracking. Moreover, the design concepts of the present invention can be applied to improve many existing MRI surface or wrap-around surface coils for imaging of other parts of human body. Also, although described with respect to imaging the human body, the coils of the present invention can be used to image any animal or other body, and the term "body" or "patient" as used herein should not be interpreted as limited to a human body or patient.

Detailed Description Text (21):
Thus, as described above ladder resonance quadrature coils constructed according to the present invention provide the superior SNR of quadrature coils with the uniformity of field obtainable with ladder resonance coils, in a single circuit design easy to construct and use. As used herein, the term "single-circuit" refers to a circuit connected together with physical electrical components or equivalents of physical components, as opposed to the independent coil designs of prior art quadrature coil systems wherein the coils are formed separately from one another. It

should also be noted that many different embodiments of the coil of the present invention are possible, and in particular that the type and placement of the circuit elements necessary for tuning, detuning and quadrature operation can be varied considerably and still achieve the desired operation, as known to those of skill in the art. Although specific embodiments have been illustrated and described herein, it will be appreciated by those of ordinary skill in the art that any arrangement which is calculated to achieve the same purpose may be substituted for the specific embodiment shown. This application is intended to cover any adaptations or variations of the present invention. Therefore, it is intended that this invention be limited only by the claims and the equivalents thereof.

Other Reference Publication (1):
Ballon, D., et al., "A 64 MHz Half -Birdcage Resonator for Clinical Imaging", J. of Magnetic Resonance, 90, 131-140, (1990).

Other Reference Publication (2):
Hu, X., et al., "Reduction of Field of View for Dynamic Imaging", Magnetic Resonance in Medicine, 31, No. 6, 691-694, (1994).

Other Reference Publication (3):
Mehdizadeh, M., " RF Coils for Magnetic Resonance Imaging", RF Design, 29-38, (1991).

Other Reference Publication (4):
Panych, L.P., et al., "A Dynamically Adaptive Imaging Algorithm for Wavelet -Encoded MRI", Magnetic Resonance in Medicine, 32, No. 6, 738-746, (1994).

CLAIMS:

1. A radio frequency (RF) coil construction, comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path.

3. A coil construction according to claim 1 wherein a first mode circuit path is sensitive to magnetic flux in a first orientation, and a second mode circuit path is sensitive to magnetic flux in a second orientation orthogonal to the first whereby quadrature operation is obtained.

4. A coil construction according to claim 3 wherein first and second relatively isolated signals representing orthogonal magnetic fields emitted from a body under investigation in an MRI apparatus are obtainable from the respective first and second mode circuit paths.

6. A coil construction according to claim 1 further including means for detuning the coil temporarily whereby interference with an externally generated RF field is avoided.

9. A coil construction according to claim 1 further wherein the coil has a half-birdcage shape.

10. A radio frequency (RF) coil construction, comprising a pair of first tuned ladder resonator coils each having two or more conductor paths in addition to a shared center conductor path, the ladder resonator coils sharing a common conductor pattern and symmetrical about the center circuit path, said construction providing a first mode circuit path formed to include the conductor paths of each of the ladder resonator coils, the first mode circuit path defining a second coil, the construction including means for tuning the first and second coils to a given RF and means for causing said first and second coils to operate in quadrature and generate first and second relatively isolated signals manifesting said orthogonal magnetic fields in said body.

11. A method of forming a single-circuit quadrature radio frequency (RF) coil, comprising the steps of a) combining a pair of first tuned ladder resonator coils to provide a common conductor pattern and so that they are symmetrical about a center conductor path, and b) adding to the combined coils elements for forming first and

second coils each respectively responsive in quadrature mode operation to magnetic fields orthogonal to one another.

12. An integrated radio frequency (RF) coil system comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path fixed in position relative to another coil of a different design, so that the quadrature coil and the coil of a different design can be alternately used when imaging a body under investigation.

13. A coil system according to claim 12 wherein the coil of a different design is a full birdcage coil and the quadrature coil is positioned to have a sensitive region for imaging a neck of a patient whose head is positioned in the full birdcage coil .

14. A method of MRI imaging a patient's neck comprising the step of positioning the patient's head in the birdcage coil of claim 13, and using either the birdcage coil or the quadrature coil for imaging operations.

15. An MRI apparatus comprising a magnet system for generating a steady, uniform magnetic field, a magnet system for generating magnetic gradient fields, a transmit RF coil for generating a RF magnetic alternating field and a surface coil comprising a single-circuit quadrature coil including a pair of tuned ladder resonator coils sharing a common conductor pattern symmetrical about a center conductor path.

16. A method for imaging the head, neck or heart of a patient comprising the step of positioning the coil of claim 1 adjacent thereto and using the quadrature coil in combination with an MRI apparatus to image the same.

17. A coil construction according to claim 1 further including a sensor for detecting the pulse of a patient under investigation, wherein the sensor is mechanically mounted in relation to the quadrature coil so that the pulse is obtained proximate the location of the coil when positioned adjacent the patient.

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PGPUB-DOCUMENT-NUMBER: 20030146750
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DOCUMENT-IDENTIFIER: US 20030146750 A1

TITLE: RF coil for imaging system

PUBLICATION-DATE: August 7, 2003

INVENTOR-INFORMATION:

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Vaughan, J. Thomas JR.	Stillwater	MN	US	

US-CL-CURRENT: 324/318; 707/104.1

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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2. Document ID: US 20020125888 A1

L17: Entry 2 of 17 File: PGPB Sep 12, 2002

PGPUB-DOCUMENT-NUMBER: 20020125888
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020125888 A1

TITLE: Magnetic resonance imaging apparatus

PUBLICATION-DATE: September 12, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Visser, Frederik	Eindhoven		NL	
Haans, Paulus Cornelius Hendrikus Adrianus	Eindhoven		NL	
Van Den Brink, Johan Samuel	Eindhoven		NL	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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3. Document ID: US 20020103429 A1

L17: Entry 3 of 17 File: PGPB Aug 1, 2002

PGPUB-DOCUMENT-NUMBER: 20020103429
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020103429 A1

TITLE: Methods for physiological monitoring, training, exercise and regulation

PUBLICATION-DATE: August 1, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
deCharms, R. Christopher	Moss Beach	CA	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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4. Document ID: US 20020087063 A1

L17: Entry 4 of 17 File: PGPB Jul 4, 2002

PGPUB-DOCUMENT-NUMBER: 20020087063
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020087063 A1

TITLE: New method and system for processing magnetic resonance signals to remove transient spike noise

PUBLICATION-DATE: July 4, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lou, Xiaoming	Waukesha	WI	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

5. Document ID: US 20020042563 A1

L17: Entry 5 of 17 File: PGPB Apr 11, 2002

PGPUB-DOCUMENT-NUMBER: 20020042563
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020042563 A1

TITLE: Method and apparatus for objectively measuring pain, pain treatment and other related techniques

PUBLICATION-DATE: April 11, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Becerra, Lino R.	Cambridge	MA	US	
Breiter, Hans C.	Lincoln	MA	US	
Borsook, David	Concord	MA	US	

US-CL-CURRENT: 600/407; 600/411, 600/420, 600/427, 600/431, 600/473, 600/475

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

6. Document ID: US 6529000 B2

L17: Entry 6 of 17 File: USPT Mar 4, 2003

US-PAT-NO: 6529000
DOCUMENT-IDENTIFIER: US 6529000 B2

TITLE: Method and system for processing magnetic resonance signals to remove transient spike noise

DATE-ISSUED: March 4, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lou; Xiaoming	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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7. Document ID: US 6501274 B1

L17: Entry 7 of 17 File: USPT Dec 31, 2002

US-PAT-NO: 6501274
DOCUMENT-IDENTIFIER: US 6501274 B1

TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

DATE-ISSUED: December 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ledden; Patrick	Malden	MA		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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8. Document ID: US 6289232 B1

L17: Entry 8 of 17 File: USPT Sep 11, 2001

US-PAT-NO: 6289232
DOCUMENT-IDENTIFIER: US 6289232 B1

TITLE: Coil array autocalibration MR imaging

DATE-ISSUED: September 11, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jakob; Peter M.	Brookline Village	MA		
Sodickson; Daniel K.	Cambridge	MA		
Griswold; Mark	Brookline	MA		

US-CL-CURRENT: 600/410; 324/307, 324/309, 324/318, 324/322, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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9. Document ID: US 5977770 A

L17: Entry 9 of 17 File: USPT Nov 2, 1999

US-PAT-NO: 5977770
DOCUMENT-IDENTIFIER: US 5977770 A

TITLE: MR imaging of synchronous spin motion and strain waves

DATE-ISSUED: November 2, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ehman; Richard L.	Rochester	MN		

US-CL-CURRENT: 324/318; 600/421

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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10. Document ID: US 5910728 A

L17: Entry 10 of 17 File: USPT Jun 8, 1999

US-PAT-NO: 5910728
DOCUMENT-IDENTIFIER: US 5910728 A

TITLE: Simultaneous acquisition of spatial harmonics (SMASH): ultra-fast imaging with radiofrequency coil arrays

DATE-ISSUED: June 8, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel Kevin	Cambridge	MA		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KMC
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11. Document ID: US 5898306 A

L17: Entry 11 of 17 File: USPT Apr 27, 1999

US-PAT-NO: 5898306
DOCUMENT-IDENTIFIER: US 5898306 A
**** See image for Certificate of Correction ****

TITLE: Single circuit ladder resonator quadrature surface RF coil

DATE-ISSUED: April 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Haiying	Minneapolis	MN		
Truwit; Charles L.	Wayzata	MN		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KMC
Draw Desc | Image

12. Document ID: US 5592085 A

L17: Entry 12 of 17 File: USPT Jan 7, 1997

US-PAT-NO: 5592085
DOCUMENT-IDENTIFIER: US 5592085 A
**** See image for Certificate of Correction ****

TITLE: MR imaging of synchronous spin motion and strain waves

DATE-ISSUED: January 7, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ehman; Richard L.	Rochester	MN		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KMC
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13. Document ID: US 5560360 A

L17: Entry 13 of 17 File: USPT Oct 1, 1996

US-PAT-NO: 5560360
DOCUMENT-IDENTIFIER: US 5560360 A
**** See image for Certificate of Correction ****

TITLE: Image neurography and diffusion anisotropy imaging

DATE-ISSUED: October 1, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Filler; Aaron G.	Seattle	WA		
Tsurda; Jay S.	Mercer Island	WA		
Richards; Todd L.	Seattle	WA		
Howe; Franklyn A.	London			GB2

US-CL-CURRENT: 600/408; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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14. Document ID: US 5212450 A

L17: Entry 14 of 17 File: USPT May 18, 1993

US-PAT-NO: 5212450
DOCUMENT-IDENTIFIER: US 5212450 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: May 18, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hills	PA		
Srinivasan; Ravi	Philadelphia	PA		
Carvajal; Lucas	North Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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15. Document ID: US 5202635 A

L17: Entry 15 of 17 File: USPT Apr 13, 1993

US-PAT-NO: 5202635
DOCUMENT-IDENTIFIER: US 5202635 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: April 13, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Philadelphia	PA		
Murphy-Boesch; Joseph	Lafayette Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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16. Document ID: US 5194811 A

L17: Entry 16 of 17 File: USPT Mar 16, 1993

US-PAT-NO: 5194811
DOCUMENT-IDENTIFIER: US 5194811 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: March 16, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hill	PA		
Srinivasan; Ravi	Philadelphia	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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17. Document ID: US 5050605 A

L17: Entry 17 of 17 File: USPT Sep 24, 1991

US-PAT-NO: 5050605
DOCUMENT-IDENTIFIER: US 5050605 A
**** See image for Certificate of Correction ****

TITLE: Magnetic resonance imaging antennas with spiral coils and imaging methods employing the same

DATE-ISSUED: September 24, 1991

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Eydelman; Gregory	West Hempstead	NY		
Giambalvo; Anthony	Kings Park	NY		
Damadian; Raymond V.	Woodbury	NY		

US-CL-CURRENT: 600/422; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KMC | Draw Desc | Image

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L17: Entry 7 of 17 File: USPT Dec 31, 2002

DOCUMENT-IDENTIFIER: US 6501274 B1
TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

Abstract Text (1):
A magnetic resonance imaging system comprises: a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis. The housing contains: a transmit /receive inductor system having a coil about the axis in proximity with the housing, a gradient inductor system having a coil operatively associated with the transmit /receive inductor system, and a field inductor system having a coil operatively associated with the transmit /receive inductor system. The field coil establishes a supervening field about the entire system. The gradient coil initiates perturbations in the fields and produces signals derived responsively from the perturbations. The transmit /receive coil includes a series of electrical transmission line elements paraxially distributed with respect to the axis about the subject. Each transmission line element includes an outer conductor and an inner conductor spaced radially from the outer conductor relative to the axis. The transmit/receive coil initially transmits to the subject a radio frequency energy field and responsively receives from the subject a magnetic resonance energy field. The signals produced correspond to spatial indicia derived from the subject and are presented as such by a master controller.

Parent Case Text (2):
The applicant herein claims the benefit of U.S. Provisional Patent Application No. 60/159,662, dated Oct. 15, 1999 for HIGH RESOLUTION MAGNETIC RESONANCE IMAGING SYSTEM in the name of Patrick Ledden, the applicant herein.

Brief Summary Text (3):
The present invention relates to magnetic resonance imaging and, more particularly, to a high resolution magnetic resonance imaging system and to components thereof.

Brief Summary Text (5):
Magnetic Resonance Imaging (MRI) has proven to be an enormously useful technology both for the detection and diagnosis of human disease as well as for research into the understanding of basic animal physiology. However, current MRI equipment has been limited by achievable signal-to-noise ratio (SNR) and by limitations in the ability to generate homogenous transmit fields for signal excitation, particularly at high magnetic field strengths.

Brief Summary Text (6):
For the acquisition of data from a nuclear magnetic resonance (NMR) signal, four separate components are required. First a static magnetic field must be generated by a permanent magnet generally of the superconducting type. Pursuant to quantum mechanics, the presence of the static magnetic field causes in a subject an energy difference between atomic spins aligned with and against this static magnetic field. The magnitude of the energy difference depends on a variety of factors, including strength of the magnetic field, size of the magnetic moments of individual atomic nuclei, and temperature. In general, a majority of the atomic spins will align with the static magnetic field and a higher energy minority of the atomic spins will align against it. When exposed to an oscillating magnetic field of proper frequency, such as is generated by an alternating current in a radio frequency (RF) coil, some of the lower energy spins aligned with the static magnetic field will be excited to

the higher energy state of being aligned against the field. Once the applied transmit RF magnetic field is removed, these excited spins will decay to the lower energy state of alignment with the static magnetic field. During the decay, these spins will generate their own RF magnetic field, which can be electronically detected by the same or a different RF coil and thereby be characterized. In order to determine spatial information about the quantity and properties of the atomic nuclei of the subject, a second set of coils, gradient coils, are used to perturb the static magnetic field. By generating magnetic field gradients, current in this separate set of coils spatially changes the oscillation frequency of the atomic spins by changing the frequency of the nuclear magnetic resonance (NMR) oscillation at appropriate times during transmit and receive, and spatial information regarding the atomic spins can be decoded and converted into an image. The generation and reception of the NMR signal in the RF coil and the currents in the gradient coils are controlled by a computer system which processes the information obtained and displays it on a computer screen or printed film for human interpretation.

Brief Summary Text (7):
The advantages of using NMR are several-fold. First, information can be obtained non-invasively on a wide variety of in vitro and in vivo subjects. The lack of non-ionizing radiation is particularly attractive when images are obtained from human subjects. Second, the properties of the magnetic spins are extremely sensitive to their surrounding chemical environment. This allows a great deal of information to be determined from the magnetic resonance signal, including chemical and molecular structure of a wide variety of materials as well as the chemical and structural characteristics of animal and human tissue. By obtaining spatially dependent information regarding the NMR signal, it is possible to obtain detailed images, which not only show great anatomic detail, but which also depend on the chemical properties of tissue. This provides additional image contrast, allows improved discrimination between healthy and diseased tissue, and permits researchers to obtain previously unavailable information regarding in vivo physiologic function.

Brief Summary Text (8):
Despite the multiple advantages of MRI, one major limiting factor in the usefulness of the NMR machine is the small magnitude of the NMR signal generated by a subject's nuclei themselves. This weak signal is easily obscured by the noise present in all electronic detection devices. The presence of this noise then limits the maximum achievable resolution or sensitivity of the NMR machine, specifically, its ability to resolve small anatomic details or to characterize time dependent changes in signal intensity, which are important for understanding of a subject's physiology.

Brief Summary Text (9):
In principle, one can improve the sensitivity of the NMR device by increasing the strength of the static magnetic field. While this does increase signal to noise ratio (SNR), it adds problems in terms of the interaction of high frequency magnetic fields and human tissue, leading to difficulties in achieving uniform image quality and even excitation of NMR spins. Simply increasing the magnetic field strength is a very expensive option: a 3T (3-Tesla) human size magnet costs roughly five times that of a 1.5T magnet. In general, such increased cost places a premium on maximizing SNR at a given field strength.

Brief Summary Text (10):
Most of the noise in human MRI comes from the resistance associated with conductive tissue within the human body. As this resistance is roughly proportional to volume of tissue, large coils, which couple to larger volume of tissue, inherently produce lower quality images than smaller coils. While sensitivity can be improved by making smaller coils, there is a limit to this approach in that eventually the desired body part or region of interest will not fit within the coil or field of view of the coil.

Brief Summary Text (11):
One prior art method designed to increase the field of view of small coils is to use multiple coils arranged in a "phased array" (U.S. Pat. No. 4,887,039). In this method, the images from each individual coil are processed separately and then combined in such a fashion as to maximize image quality. While this is a useful

strategy, it has certain limitations. First the individual coils need to be carefully oriented to minimize their respective coupling. Despite proper orientation, there always will be residual coupling between four or more coils limiting the maximum number of coils and consequently the gains in sensitivity. Furthermore, in the standard geometry feasible with surface coils, this arrangement still produces inhomogeneous images, which can complicate their interpretation for diagnostic purposes.

Brief Summary Text (12):
A second problem is the efficient and uniform excitation of the NMR spins. For most imaging sequences, a homogenous excitation of all spins is required. In general, this requires a larger coil, which then reduces the sensitivity of the system. One commonly used technique is to use a larger coil, optimized for transmit with a second coil specialized for receive. However such systems, as presently implemented, suffer from several disadvantages, particularly when used in high field systems. One disadvantage of current volume transmit coils is the inability to control the field to compensate for variations in patient size and position. While, in principle these variations can be accomplished by manually tuning the coil (see J. Thomas Vaughan, Hoby P. Hetherington, Joe O. Out, Jullie W. Pan, Gerald M. Pohost, "High Frequency Volume Coils for Clinical NMR Imaging and Spectroscopy", Magnetic Resonance in Medicine 32:206-218 (1994)) or by using electromechanical relays to switch in additional reactive circuit elements, such methods are time consuming and subject to the variability of mechanical connections.

Brief Summary Text (13):
Conventional MRI coils come in two basic categories. (1) The simpler, the surface coil, consists of one or more conductive loops. Additional reactive circuit components, such as capacitors and inductors, are used to tune the coil and couple energy to or from it to the rest of the NMR system. Importantly, active circuit elements, such as PIN diodes, can be added to allow specialization of coil function for receive or transmit. (2) Volume coils, such as birdcage coils, consist of one or more large surface coils oriented in such a fashion as to produce a homogenous magnetic field. While such coils are in common use, the large size of these coils makes them poor receivers of NMR signal. This difficulty can be overcome by using PIN diodes to "detune" the volume coil for use with a more sensitive surface coil receiver.

Brief Summary Text (14):
In particular, at high fields, the use of volume coils becomes increasing problematic. The large size of these coils required to enclose a useful area of human anatomy, such as the torso or head, leads to them becoming efficient radiators of electromagnetic energy. Moreover, the interaction of large volume coils with tissue at high frequencies leads to non-uniform magnetic fields within human tissue complicating the ability to obtain uniform spin excitation.

Brief Summary Text (15):
The following U.S. Pat. No. 5,557,247 to Vaughn, U.S. Pat. No. 4,751,464 to Bridges, U.S. Pat. No. 4,746,866 to Roschmann and U.S. Pat. No. 4,506,224 to Krause, disclose volume coils based on cavity resonators. Conductive segments within the cavity interact to form a resonant structure. While this coil can offer improved efficiency over a conventional volume coil, several disadvantages exist. First, the structure being closed can give a subject a sense of claustrophobia and make it difficult to present visual stimulation for research purposes. Second, the closed shielded nature of the coil makes it difficult to specialize for the use of transmit or receive purposes. If circuit elements are added to detune the coil, the outer cavity shield will interact with smaller coils placed with the larger cavity, impairing their performance. Additionally, the cavity shield prevents the use of the coil for specialization as a smaller coil to use with receive only function or as its use as a phased array.

Brief Summary Text (17):
The present invention is an improved NMR coil design based on the use of transmission line segments rather than conventional inductive coil elements. The use of transmission lines has several benefits. Transmission lines have a concentration of electromagnetic fields between their elements. By adjusting the distance between

these conductive elements, interaction of the magnetic fields of the transmission line with an external sample can be controlled and optimized for NMR signal generation and/or detection. The presence of two conductors also decreases the inductance of each conductor. This minimizes the electric fields associated with the conductors, which is advantageous since these electric fields can be associated with dielectric tissue losses which decrease coil efficiency and sensitivity. Moreover, the inherent shielded nature of transmission lines decreases the radiation of electromagnetic energy from the NMR coil, improving coil efficiency and sensitivity over conventional NMR coil design. The shielded nature of a transmission line also decreases the interaction or coupling between coil elements. This can be advantageous since under proper conditions, coil elements can operate with minimal interaction. This allows a large single coil structure to operate as multiple smaller individual coils. With proper combinations, these separate coils can be combined in such a way to optimize NMR signal generation and/or reception. In particular, by combining signals from individual coil elements, spatial information may be decoded regarding the NMR signal, increasing the sensitivity and speed of data acquisition for both high field and low field NMR systems.

Brief Summary Text (18):
The coil consists of N transmission line segments distributed in a circular, elliptical, or other geometrical arrangement. Each transmission line element is comprised of two or more individual conductors with or without additional lumped or distributed capacitive or inductive circuit components. In general, each transmission line element couples to the others through mutual inductance and capacitive coupling. Additional lumped or distributed inductive or capacitive elements may be placed between the transmission line segments to alter this coupling. The combined influences of the interaction between these elements gives rise to frequency dependent relations between the currents and voltages present on individual transmission line elements. By changing the individual circuit components and transmission line geometry, a given current distribution can be obtained on the transmission line elements at a given frequency. The magnetic field arising from the currents on each element add through superposition and create a given magnetic field configuration for use in either or both the generation and detection of the NMR signal.

Brief Summary Text (19):
In particular, with placement of properly valued reactive components between individual transmission line elements, mutual coupling between elements can be minimized. This allows the resonant structure of the N transmission line segment to become degenerate and allows the currents on each element to be relatively independent. This has the advantage for NMR signal generation in that the currents on each element can be individually controlled at will to generate a excitation field of a desired spatial and phase characteristic. Additionally in such an degenerate mode arrangement, received signals from each element are independent and can then be combined in such a way to optimize image homogeneity, sensitivity, or other desired parameters.

Brief Summary Text (20):
In order for the transmission line structure to be useful, energy needs to be transferred into the coil during signal generation and out of the coil during signal reception. This can be accomplished by inductively or capacitively coupling one or more circuit elements to one or more RF power amplifiers and/or RF receivers. This coupling can be adjusted to allow an arbitrary impedance of such equipment to be matched to the currents and voltages found in the transmission line structure. In particular, the phases of the current in two or more transmission line elements can be offset as to create elliptically polarized magnetic fields for improved efficiency in the generation and/or detection of the nuclear magnetic resonance signal.

Brief Summary Text (21):
In addition to passive components, active circuit elements such as diodes (either regular or PIN) can be added to this structure. With diodes, the tuning of individual transmission line elements or their mutual coupling can be changed in order to modify the current distribution and element impedance of the transmission line segments. When used with one or more additional coils (which may be a

combination of transmission line structures or conventional NMR coils), these diodes can be arranged so that during transmit or receive functions, one coil has a desired magnetic field configuration while the other coil presents a high impedance so as not to interfere with the magnetic fields of the first coil. In this manner, each of the two or more coils can be optimized for either transmit or receive, resulting in improved generation and detection of the NMR signal.

Brief Summary Text (22):
Other active circuit elements can be added to the transmission line structure such as vacuum tubes or transistors (including but not limited to conventional bipolar transistors, field effect transistors, gallium arsenide field effect transistors, high electron mobility transistors, pseudomorphic high electron mobility transistors, or heterojunction bipolar transistors). These transistors can be used to provide amplification of either the transmit energy needed in the generation of the nuclear magnetic signal or the small magnitude received energy from the NMR spin decay. In this way, signal losses arising from matching circuits and connecting cables are minimized, leading to improved coil efficiency. If the coil is designed for both transmit and receive functions, diodes may be included to change the coupling between these active amplifier circuits and individual transmission line elements. In this manner, transistors designed for low-noise signal amplification are not damaged by the high element currents during the transmit function and transistors circuits designed for power amplification do not add noise during signal reception.

Brief Summary Text (23):
The addition of active vacuum tube or transistor circuits can provide additional advantages. With proper design, these circuits can present impedance mismatches to the transmission line structure while simultaneously preserving adequate amplifier function. These impedance mismatches can be used to change or minimize coupling between individual transmission line elements, allowing the elements to be decoupled and be relatively independent of each other. During transmit, this has the advantage that individual element currents can be changed electronically in magnitude or phase so as to modify the desired magnetic field for optimal transmit excitation without requiring change or variation of passive circuit elements. This is particularly advantageous at high frequencies where dielectric resonances in human tissue require non-uniform magnetic fields for uniform spin excitation. Additionally, during receive, decoupling of the currents on transmission line elements allows each element to function as a separate signal detector. By combining the signals from these elements electronically, either directly after amplification or at a later stage such as after image reconstruction, these signals can be added in a way such that sensitivity is maximized for one or more areas of interest. In particular, the spatially dependent information from each element can be combined after image reconstruction in such a manner that sensitivity is maximized at each point in an image.

Brief Summary Text (24):
Moreover, the geometric arrangement of the individual transmission line elements can be used to decode spatial information regarding the detected NMR signal. By decoding spatial information from individual coil elements, the steps required for the acquisition of an NMR image can be reduced, allowing the imaging process to be completed in less time.

Brief Summary Text (25):
The illustrated embodiments of the present invention demonstrate an actively decoupled transmission line resonator for use as a transmit coil in conjunction with surface coil receivers, as well as use of a transmission line structure as a receive array coil.

Drawing Description Text (4):
FIG. 1a illustrates a partial assembly of a transmission line coil, with one line element and the rear end plate;

Drawing Description Text (5):
FIG. 2 illustrates a complete transmission line assembly of the type shown in FIG. 1;

Drawing Description Text (6):
FIG. 3 illustrates a frame for the transmission line assembly of FIG. 2;

Drawing Description Text (7):
FIG. 4 illustrates a completed coil incorporating the transmission line assembly of FIG. 2 and the frame of FIG. 3;

Drawing Description Text (8):
FIG. 5 is an end view of a detunable head transmit coil embodying the present invention;

Drawing Description Text (9):
FIG. 6 is a side view of the detunable head transmit coil of FIG. 5;

Drawing Description Text (13):
FIG. 10 is an electrical schematic of a receive-only circuit in accordance with the present invention;

Drawing Description Text (14):
FIG. 11 illustrates an image produced by a head transmit coil for both transmit and receive in accordance with the present invention;

Drawing Description Text (15):
FIG. 12 illustrates an image produced by a head transmit coil for transmit and a dual loop coil for receive in accordance with the present invention;

Drawing Description Text (16):
FIG. 13 illustrates a human brain image produced by a head transmit coil used for both transmit and receive;

Drawing Description Text (17):
FIG. 14 illustrates a human brain image produced by a head transmit coil used for transmit and a loop coil for receive;

Drawing Description Text (18):
FIGS. 15, 16 and 17 illustrate image intensity corrected images produced by receive-only coils;

Drawing Description Text (19):
FIG. 18 illustrates a power amplifier directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (20):
FIG. 19 illustrates a low-noise transistor amplifier directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (21):
FIG. 20 illustrates a combination of active transmit and receive circuits directly connected to an individual transmission line element in accordance with the present invention;

Drawing Description Text (22):
FIG. 21 illustrates an overall system diagram in connection with a transmission line coil used for both transmit and receive in accordance with the present invention;

Drawing Description Text (23):
FIG. 22 illustrates an overall system diagram in connection with separate transmit and receive coils in accordance with the present invention;

Drawing Description Text (24):
FIG. 23 illustrates an overall system diagram in connection with a transmission line coil used in array mode in accordance with the present invention;

Drawing Description Text (25):

FIG. 24 illustrates images obtained from four element transmission line coil operated in the receive array mode whereby low noise preamplifiers detune the interactions between elements allowing each transmission line element to function independently ;

Drawing Description Text (27):
FIG. 26 illustrates human images obtained from an elliptic four element transmission line coil operated in the receive array mode whereby a combination of capacitors and low noise preamplifiers detune the interactions between elements allowing each transmission line element to function independently ; and

Drawing Description Text (28):
FIG. 27 illustrates the sum-of-squares combination of the images in FIG. 26 to create a high sensitivity homogenous image of the human brain.

Detailed Description Text (3):
Generally, the illustrated system of the present invention includes a housing 11 within which are a field coil 13, X, Y, Z gradient coils 15 and a transmit /receive coil 17. Field coil 13 is energized by a field coil controller 19. Gradient coils 15 are controlled by a gradient coil controller 21. Transmit /receive coil 17 is controlled by a transmit /receive coil controller 23. Field coil controller 19, gradient coil controller 21, and transmit /receive controller 23 are managed by a system controller 25. A carriage 27, upon which a subject reclines, is reciprocable into and out of the region 29 within transmit /receive coil 17. Region 29 is a medical diagnostic chamber within which the subject is internally imaged pursuant to the present invention.

Detailed Description Text (4):
As shown in FIGS. 1a to 4, the coil at the heart of the present invention, is a transmit and/or receive coil that includes transmission line elements distributed in a circular, elliptical, or other geometrical arrangement. FIG. 1a, for simplicity, shows a single transmission line element 12a (collectively, 12) mounted to a circular end cap 14 with slots 22. The present invention contemplates that the end cap 14 may have another shape, such as a dome, or may be absent altogether. FIG. 2 shows a complete transmission line assembly 16 with all 16 transmission line elements 12. Note that there are gaps 24, corresponding the end cap slots 22, between each of the elements 12. FIG. 3 shows a mechanical frame 18 into which the transmission line assembly 16 is fixed. FIG. 4 shows an entire coil structure 10 including an outside case or housing 20.

Detailed Description Text (5):
Each transmission line element 12 comprises two or more individual conductors 26, 28 with or without additional lumped or distributed capacitive or inductive circuit components. Each transmission line element 12 couples to the others through mutual inductance and capacitive coupling. Additional lumped or distributed inductive or capacitive elements, in various embodiments, are placed between transmission line segments (groups of transmission line elements 12) to alter this coupling. The combined influences of the interaction between the elements 12 give rise to frequency-dependent relations between the currents and voltages present on individual transmission line elements 12, as shown in the Kirchoff circuit relationship between transmission line currents and voltages: ##EQU1##

Detailed Description Text (6):
where Xelem.sub.j is the complex impedance of transmission line element n at frequency .omega., Xm.sub.i,j is the complex impedance associated with the coupling between elements i and j, I.sub.n is the current through element n, and V.sub.n is the voltage across element n. By changing the individual circuit components and transmission line geometry, a given current distribution can be obtained on the transmission line elements 12 at a given frequency. The magnetic field arising from the currents on each element 12 add through superposition to create a given magnetic field configuration for use in either or both the generation and detection of an NMR signal .

Detailed Description Text (7):
In order for the transmission line structure to be useful, energy must be

transferred into the coil 10 during signal generation and out of the coil 10 during signal reception. This can be accomplished by inductively or capacitively coupling one or more elements 12 to one or more RF power amplifiers and/or RF receivers . The coupling can be adjusted to allow an arbitrary impedance of such equipment to be matched to the currents and voltages found in the coil 10. In particular, the phases of the current in two or more transmission line elements 12 can be offset so as to create circularly or elliptically polarized magnetic fields for improved efficiency in the generation and/or detection of the NMR signal.

Detailed Description Text (8):
In addition to passive components, in various embodiments, active circuit elements such as diodes (either regular or PIN) are added to the circuit. With diodes, the tuning of individual transmission line elements 12 or their mutual coupling can be changed in order to modify the current distribution and element impedance of the transmission line elements 12. When used with one or more additional coils 10 (which may be a combination of transmission line structures or conventional NMR coils), these diodes can be arranged so that, during transmit or receive functions, one coil 10 has a desired magnetic field configuration while another coil 10 presents a high impedance, so as not to interfere with the magnetic fields of the first coil 10. In this manner, each of the two or more coils 10 can be optimized for either transmit or receive, resulting in improved generation and detection of the NMR signal.

Detailed Description Text (9):
In addition to diodes, in various embodiments, other active circuit elements are added to the transmission line structure 16. These include vacuum tubes or transistors (including but not limited to conventional bipolar transistors, field effect transistors, gallium arsenide field effect transistors, high electron mobility transistors, or heterojunction bipolar transistors). Transistors can be used to provide amplification of either the transmit energy needed in the generation of the NMR signal or the small received energy from the NMR spin decay. In this way, signal losses arising from matching circuits and connecting cables are minimized, leading to improved coil efficiency. If the coil 10 is designed for both transmit and receive functions, diodes may be included to change the coupling between the active amplifier circuits and individual transmission line elements 12. In this manner, transistors designed for low-noise signal amplification are not damaged by the high element currents during the transmit function, and transistor circuits designed for power amplification do not add noise during signal reception.

Detailed Description Text (10):
The addition of active vacuum tube or transistor circuits can provide additional advantages. Pursuant to the present invention, these circuits can present impedance mismatches to the transmission line structure 16 while simultaneously preserving adequate amplifier function. Impedance mismatches can be used to change or minimize coupling between individual transmission line elements 12 allowing the elements' currents to be decoupled and relatively independent of each other. During transmit, this has the advantage that individual element currents can be changed electronically in magnitude or phase so as to modify the desired magnetic field for optimal transmit excitation without requiring change or variation of passive circuit elements. This is particularly advantageous at high frequencies where dielectric resonance in human tissue require non-uniform magnetic fields for uniform spin excitation.

Detailed Description Text (11):
Additionally, during receive, decoupling of the currents on transmission line elements 12 allows each element 12 to function as a separate signal detector. By combining the signals from these elements 12 electronically, either directly after amplification or at a later stage such as after image reconstruction, these signals can be added in such a way that sensitivity is maximized for one or more areas of interest. In particular, the spatially dependent information from each element 12 can be combined after image reconstruction in such a manner that sensitivity is maximized at each point of an image. Moreover, the geometric arrangement of the individual transmission line elements 12 can be used to decode spatial information regarding the detected NMR signal. By decoding spatial information from individual coil elements 12, the steps required for the acquisition of an NMR image can be reduced, allowing the imaging process to be completed more quickly.

Detailed Description Text (13):
FIGS. 5 and 6 show the geometry of a detunable volume transmit coil 40 constructed in accordance with the present invention. The volume coil 40 is a flat element, shielded transmission line design that is analogous to a previously described design utilizing coaxial elements. (See U.S. Pat. No. 4,746,866 to Roschman and U.S. Pat. No. 5,557,247 to Vaughan.) The coil 40 incorporates an end-capped structure that decreases radiation losses. A conductive cavity wall 42 is divided into 12 outer conductors 44 by regularly spaced longitudinal slots 46. The slots 46 minimize eddy currents when used in echo planar imaging (EPI). In one configuration, the diameter of the coil 40 is 37.5 cm and the axial length is 20 cm. Twelve flat copper inner conductors 50, each with a width of 2.5 cm, are located 1.75 cm inwardly from the cavity wall 42, as at 52, and centered between the slots 46. The inner conductors 50 are tuned using nonmagnetic chip capacitors and nonmagnetic tuning capacitors. Two elements 12a, 12d, located 90.degree. from each other, are matched to 50.OMEGA. using lumped element quarter-wave transformers. These outputs then are driven through a quadrature coupler.

Detailed Description Text (14):
In one configuration, shown in FIG. 7, detuning is accomplished with a diode 60. The circuit shown in FIG. 8 utilizes a shunt diode configuration similar to that described in Ledden, P. J., Wald, L. L., Vaughan, J. T., "Volume Coil Transmit Surface Coil Receive System for Brain Imaging at 3T", Proceedings of the International Society of Magnetic Resonance in Medicine, p. 168 (1999). In this arrangement, a diode 62 is placed across the tuning capacitor 64 at the posterior end of every element 12. During transmit, the diodes 62 are back biased and have a high impedance allowing normal tuned coil operation. During receive, the diodes 62 are forward biased, shorting the tuning capacitors 64 and detuning the coil 10. Bias voltage is applied to each diode 62 through high impedance RF chokes 66 which have an RF impedance of greater than 1 K.OMEGA. at 127.8 MHz. The diode bias voltage is provided by a coaxial cable 68 separate from the RF connections and is controlled by a 5 V digital signal from a scanner.

Detailed Description Text (15):
In another configuration, shown in FIG. 9, detuning utilizes a lumped element quarter wave line 76 between the diode 72 and the tuning capacitor 74. In this arrangement, the diode 72 is forward biased during transmit and shorts the quarter wave circuit 76. This causes the quarter wave circuit 76 to present a high impedance across the tuning capacitor 74 allowing normal coil resonance. During receive, the diode 72 is back biased and causes the quarter wave circuit 76 to short the tuning capacitor 74, thereby detuning the coil 10. Unlike the configuration in FIG. 9, high negative bias voltages are not required, since during receive, the RF voltages in the coil 10 are very small. This eliminates the need for a high voltage bias supply and driver resulting in improved operator and patient safety.

Detailed Description Text (16):
Two different geometries of receive-only surface coils are presented herein. The first coil consists of a quadrature surface coil 70 comprising two 9-cm loops 72, as in FIG. 10. Of similar overall design, the second coil consists of two 12 cm.times.20 cm curved rectangular loops, also combined in quadrature. Each coil is matched to 50.OMEGA. using a standard balun drive circuit 74. Detuning during transmit is accomplished by placing a PIN diode 76 across the balun 74. In the conductive state, this diode 76 shorts the balun 74, causing the coil 70 to double tune with a null at the 128 MHz.

Detailed Description Text (17):
All electrical impedance measurements in the aforementioned embodiments were made with a network analyzer. The isolation produced by the diode detuning of the transmit coil was determined by the change in radio frequency transmission between two untuned 2.5-cm-diameter probe coils loosely coupled to the volume coil. The isolation was taken as the difference in radio frequency insertion loss in decibels at 127.8 MHz between the tuned and detuned states. The two probe coils were physically separated and made electrically orthogonal to minimize their inductive coupling. A similar method was used to measure the degree of detuning obtained by the active PIN diode trap structure on the receive surface coils.

Detailed Description Text (19):
All studies were performed using a 3T system incorporating an 80-cm bore magnet, and a resonant gradient coil for EPI. Coil SNR was calculated by dividing the image intensity by the standard deviation of the background noise. Transmit efficiency was compared to the standard commercial 16-rung birdcage coil (28 cm diameter and 30 cm length) provided with the system by comparing the transmit gain required for 90.degree. spin echo excitation. All human studies were conducted with Institutional Review Board (IRB) approval.

Detailed Description Text (21):
Measurements of the detuning of the transmit coil were as follows. Greater than 40 dB of isolation was achieved between the tuned and detuned states using either of the diode detuning methods. With careful adjustment, the PIN diode trap circuit on the receive coils also provided greater than 40 dB of isolation in the detuned state. Less than 100 kHz change in loaded resonant frequency occurred when either receive-only surface coil was placed within the detuned transmit coil.

Detailed Description Text (23):
The transmit power required for a 90.degree. pulse excitation for the transmission line resonator was compared to a standard commercial birdcage coil. Despite its larger size, the transmission line resonator without detuning circuitry had approximately 10% greater efficiency than did the birdcage design. Some loss of coil efficiency occurred with either detuning circuit. Addition of the direct shunt diode configuration reduced the transmit efficiency by an amount dependent on the reverse bias voltage applied. Conversely the quarter wave diode short configuration required only enough bias voltage to forward bias the diode, but coil efficiency depended on bias current. With the direct shunt diode circuit, coil efficiency was reduced approximately 0.5 db with -250 V diode back bias. In comparison, the quarter wave diode short circuit reduced coil efficiency 75 db with 200 mA forward bias current.

Detailed Description Text (24):
FIGS. 11 and 12 show phantom images taken with the head transmit coil system. FIG. 11 shows the volume coil transmitter being used both for transmit and receive. With the relatively large size of the head transmit coil, highly uniform transmit excitation was achieved. FIG. 12 shows the results when the detunable volume coil was used for transmit and a dual 9-cm loop pair was used for receive. Image intensity decreased smoothly as a function of distance from the receive coil elements, indicating the absence of surface coil focusing of the transmit fields and good detuning of the transmit coil during receive.

Detailed Description Text (25):
FIGS. 13 and 14 show other results obtained with a human subject. In FIG. 13, the transmit coil was used for both transmit and receive. The relatively large size of the transmit coil results in a uniform transmit field over the human brain. Image SNR was approximately 10% greater than with the birdcage head coil. FIG. 14 shows the results obtained when the detunable volume coil was used for transmit and the dual 9-cm loop pair was used for receive. As with the phantom image, the surface coil receivers markedly increased local SNR. In comparison to the birdcage head coil, the combination of the head transmit coil and receive-only 9-cm loop pair provided up to 500% improvement in cortical SNR.

Detailed Description Text (26):
FIGS. 15, 16 and 17 show intensity corrected images obtained with the receive-only coils. In comparison with the smaller 9-cm loop pair, the larger rectangular quadrature coil provided up to a 350% increase in occipital SNR with broad coverage including most of the posterior half of the brain. The increased SNR with both of these receive-only coils allowed images with 400-.mu.m in-plane resolution and 1.5-mm slices to be obtained with a minimum of signal averages. As seen in the magnified images, the combination of high in-plane resolution and thin slice thickness allows subtle cortical details to be

Detailed Description Text (27):
These results demonstrate the feasibility of a volume coil transmit, surface coil receive system for brain imaging at 3T. Despite the high frequency and close

proximity to the surface coil, adequate isolation was achieved between the detuned transmit resonator and the surface coil receiver during both transmit and reception. The receive-only surface coil provided a marked increase in local SNR.

Detailed Description Text (29):
For the small receive coil employed in this study, designs derived from 1.5T applications are found to be satisfactory with increased distribution of the coil tuning capacitance. The capacitance is found to require 3 to 4 times as many distributed capacitors than at 1.5T in order to minimize load dependent frequency change. When using a resonant trap detuning circuit, stable coil resonance is important not just for optimal system noise figure but also for good isolation in the detuned state. Frequency dependent load changes are also minimized by the use of a balun drive, which symmetrized the electric fields of the coil.

Detailed Description Text (30):
The phantom and human images demonstrate the feasibility of PIN diode detuning of high frequency transmission line resonators for use with surface coil receivers at 3T. The use of small local surface coil receivers allow improved SNR for a wide variety of brain imaging applications at 3T and enable full utilization of the increased sensitivity of high field MR systems.

Detailed Description Text (31):
As indicated above, the present invention contemplates the connection of active transistor amplifiers directly to one or more of the transmission line elements. Thus, FIG. 18 shows a power transistor amplifier 80 directly connected to an individual transmission line element 12. FIG. 19 shows a low-noise transistor amplifier 82 directly connected to an individual transmission line element 12. FIG. 20 shows a combination of an active transmit circuit 84 and active receive circuit 86 directly connected to an individual transmission line element 12 with PIN diode switching. Transistor and PIN diode DC connections are omitted for clarity.

Detailed Description Text (33):
FIG. 21 is a diagram of a system 100 that uses a transmission line coil used for both transmit and receive. As shown, one or more coil elements 102 are connected through matching circuits directly or through a signal combiner, such as a quadrature combiner, to a transmit/receive (T/R) switch 104. During NMR signal generation, the T/R switch 104 connects the transmission line coil 102 to an RF generator 108 through a power amplifier 106. During NMR signal detection, the T/R switch 104 connects the coil 102 to a signal receiver 112 through a low-noise preamplifier 110. In conjunction with properly timed magnetic field gradients coils 114, the system controller 116 acquires data and processes it into an image or other useful form.

Detailed Description Text (34):
FIG. 22 is a diagram of a system 120 that uses separate transmit coils 122 and receive coils 124. As shown, an RF generator 126 is connected through an RF power amplifier 128 and T/R switch 130 to one or more transmission line coil elements 122. During transmit, PIN diodes in the coils 122 are biased to allow normal tuned coil operation, while PIN diode circuits detune the receive coil 124. During NMR signal detection, the T/R switch 130 connects the receive coil 124 to a signal receiver 132 through a low-noise preamplifier 134. PIN diodes detune the transmit coil 122, while the receive coils 124 are biased for normal tuned operation. In conjunction with properly timed magnetic field gradient coils 136, the system controller 138 acquires data and processes it into an image or other useful form.

Detailed Description Text (35):
FIG. 23 is a diagram of a system 150 that uses a transmission line coil 152 in array mode. As shown, one or more individual transmission line elements 154 are connected through built-in transistor amplifiers and T/R switches to separate NMR receivers 158 and RF generators 156. During NMR signal generation, each T/R switch connects a transmission line coil element 154 to a generator 156. During NMR signal detection, each T/R switch connects a coil element to a separate signal receiver 158. In conjunction with properly timed magnetic field gradient coils 160, the system controller 162 acquires data and processes it into an image or other useful form. By using separate RF signal generators 156 and receivers 158 for each coil element 154,

the signals from individual elements 154 can be optimally controlled and processed for maximum advantage.

Detailed Description Text (36):
FIG. 24 shows results obtained from such array system. In this case, separate preamplifier /receiver channels on a General Electric (GE) 1.5T MRI scanner are connected to each of four separate transmission line elements arranged in a cylindrically symmetric fashion similar to the 16 element array shown in FIGS. 1a -4 the impedance mismatch between the low-noise preamplifiers detunes the mutual inductive coupling between elements and allows the currents on each element to be independent . A PIN diode circuit detunes each transmission line element during transmit to allow the larger coil to generate a highly uniform spin excitation field. The images from each transmission line element are independent and can be processed separately to create an image of desired spatial sensitivity .

Detailed Description Text (37):
FIG. 25 shows the combination of each separate receive channel combined either to create a homogenous image (right) or a gradient mode image (left). Depending on the frequency of operation, a combination of these two images can be used to correct for image intensity variations caused by dielectric resonances or other high-field image artifacts.

Detailed Description Text (38):
FIG. 26 shows the results obtained with a transmission line array coil with a human volunteer. In this case, the four transmission line elements were arranged in an elliptical fashion to more closely fit the geometry of the human head. A combination of capacitors between elements and the low-noise preamplifiers detuned the coupling between elements allowing each to operate independently . As in FIG. 24, a separate preamplifier /receiver was connected to each transmission line element. A PIN diode detuning circuitry allowed the use of a highly homogenous transmit coil to provide uniform spin excitation. As seen in the images, each transmission line element operates independently providing a high sensitivity image of a portion of the human brain.

Detailed Description Text (39):
FIG. 27 shows a sum-of-squares recombination of the images in FIG. 26. By appropriately combining multiple high -sensitivity images obtained from individual transmission line elements, a homogenous image is obtained which has higher local sensitivity than could be obtained with combining signals from each element with the fixed amplitude and phase relationships as found in FIG. 13.

Detailed Description Text (40):
Thus it has been shown and described an improved NMR coil design based on the use of transmission line elements which satisfies the objects set forth above.

Detailed Description Text (41):
Since certain changes may be made in the present disclosure without departing from the scope of the present invention, it is intended that all matter described in the foregoing specification and shown in the accompanying drawings be interpreted as illustrative and not in a limiting sense .

Other Reference Publication (1):
Ledden et al., "Volume Coil Transmit Surface Coil Receive System for Brain Imaging at 3T", Proceedings of the International Society of Magnetic Resonance in Medicine, p. 168 (1999).

CLAIMS:

1. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a transmit /receive inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit /receive inductor system; (d) a static magnetic field inductor system operatively associated with said transmit /receive inductor system; (e) said transmit /receive inductor system constituting a coil having an outer surface about said axis and including a

series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface; (f) said coil initially transmitting to said subject fields of radio frequency energy as a transmit signal, and responsively receiving from said subject fields of magnetic resonance energy as a receive signal; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

2. The magnetic resonance imaging system of claim 1 wherein said coil establishes concentrations of electromagnetic fields among said transmission line segments.

3. The magnetic resonance imaging system of claim 2 wherein, by adjusting the distance between said transmission line segments, the interaction of the magnetic fields of said transmission line segments with an external sample can be controlled and optimized for nuclear magnetic resonance signal generation and/or detection.

4. The magnetic resonance imaging system of claim 1 wherein said plural transmission line segments decrease the inductance of each line segment and minimize the electric fields associated therewith, whereby dielectric tissue losses is said subject are reduced.

5. The magnetic resonance imaging system of claim 1 wherein said plural transmission line segments have inherent shielding, whereby coupling between said transmission line segments is controlled.

6. The magnetic resonance imaging system of claim 1 wherein said plural line segments are combined to optimize NMR signal generation and/or reception.

7. The magnetic resonance imaging system of claim 1 wherein signals form said plural line segments are combined to decode spatial information derived from the NMR signal, thereby to increase the sensitivity and speed of data acquisition.

8. The magnetic resonance imaging system of claim 1 wherein said inductor consists of N transmission line segments arranged in a geometric pattern in which said line segments are substantially equidistant from each other.

9. The magnetic resonance imaging system of claim 1 wherein said geometric pattern is circular or elliptical.

10. The magnetic resonance imaging system of claim 1 wherein said geometric pattern is flat or curved.

11. The magnetic resonance imaging system of claim 1 wherein each of said transmission line segments includes at least two individual conductors together with additional lumped or distributed capacitive or inductive circuit components.

12. The magnetic resonance imaging system of claim 1 wherein each transmission line element couples to the others through mutual inductance and capacitive coupling.

13. The magnetic resonance imaging system of claim 1 wherein distributed impedance elements are connected between certain of said transmission line segments to alter the coupling therebetween.

14. The magnetic resonance imaging system of claim 1 wherein impedance elements are connected between said transmission line segments to establish interactions that establish frequency dependent relations between the currents and voltages present on certain of said transmission line segments.

15. The magnetic resonance imaging system of claim 1 wherein a given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

16. The magnetic resonance imaging system of claim 1 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use in either or both the generation and detection of the NMR signal.

17. The magnetic resonance imaging system of claim 1 including RF power amplifiers and/or RF receivers coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal and out of said coil during the reception of said receive signal.

18. The magnetic resonance imaging system of claim 1 including at least an RF power amplifier reactively coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal, the impedance of said RF power amplifier and the impedance of said one of said transmission line elements being matched.

19. The magnetic resonance imaging system of claim 1 including at least an RF receiver reactively coupled to at least one of said transmission line elements for transferring energy from said coil during the reception of said receive signal, the impedance of said RF receiver and the impedance of said one of said transmission line elements being matched.

20. The magnetic resonance imaging system of claim 1 wherein the phases of the current in a plurality of said transmission line segments are offset so as to create an elliptically polarized magnetic field for generating and/or detecting nuclear magnetic resonance signals.

21. The magnetic resonance imaging system of claim 17 including a plurality of diodes operatively connected to a plurality of said transmission line segments for tuning the coupling between said transmission line segments and said RF amplifiers and receivers.

22. The magnetic resonance imaging system of claim 1 including reactive coupling elements between one or more transmission line elements to allow the currents on each transmission line element to be relatively independent.

23. The magnetic resonance imaging system of claim 1 with individual preamplifiers connected to each transmission line element with impedance mismatches designed to allow each transmission line element to operate independently allowing the signals from each transmission line element to be combined either before or after image reconstruction for optimal image reception.

24. The magnetic resonance imaging system of claim 1 with individual preamplifier /receivers connected to each transmission line element with the independent information obtained from individual transmission line elements being used to decode spatial information regarding said subject.

25. The magnetic resonance imaging system of claim 1, with individual power amplifiers connected to each transmission line element with impedance mismatches designed to allow the current of each transmission line element to be independently controlled allowing a transmit field of desired spatial intensity and phase to be generated.

26. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber with a static homogenous magnetic field for a subject therewithin lying along an axis; (b) a plurality of transmit /receive inductor systems about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit /receive inductor systems; (d) a static magnetic field inductor system operatively associated with said transmit /receive inductor systems; (e) at least one of said transmit /receive inductor systems constituting a coil having an outer surface about said axis and including a series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor

being spaced from said outer conductor in a direction perpendicular to said outer surface; (f) each said coil selectively transmitting to said subject fields of radio frequency energy, and selectively receiving from said subject fields of magnetic resonance energy; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

27. The magnetic resonance imaging system of claim 26, wherein one of said coils is a conventional loop inductor.

28. The magnetic resonance imaging system of claim 26, wherein one of said coils is a conventional loop inductor which is detuned during transmit function, said transmit function being performed by a transmission line coil which is detuned during receive.

29. The magnetic resonance imaging system of claim 26, wherein one of said coils is a phased array of conventional loop inductors.

30. The magnetic resonance imaging system of claim 26, wherein one of said coils is a phased array of conventional loop inductors which are detuned during transmit function, said transmit function being performed by a transmission line coil which is detuned during receive function.

31. The magnetic resonance imaging system of claim 26, wherein one of said coils is an array of said transmission line elements each operated independently with individual preamplifiers /receivers.

32. The magnetic resonance imaging system of claim 26, wherein one of said coils is an array of said transmission line elements each operated independently with individual preamplifiers /receivers, said array being detuned during system transmit function.

33. The magnetic resonance imaging system of claim 26, wherein said system includes at least two coils, one of said coils being a transmit coil and the other of said coils being a receive coil.

34. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a transmit inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said transmit inductor system; (d) a static magnetic field inductor system operatively associated with said transmit inductor system; (e) said receive inductor system constituting a coil having an outer surface about said axis and including a series of electrical transmission fine elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface, said coil including a means for detuning said coil to prevent disturbance of the transmit fields generated by a separate transmit inductor system; (f) said coil initially transmitting to said subject fields of radio frequency energy as a transmit signal; (g) said gradient inductor system initiating perturbations in said fields.

35. The magnetic resonance imaging system of claim 34 wherein said coil establishes concentrations of transmit electromagnetic fields among said transmission line elements.

36. The magnetic resonance imaging system of claim 34 wherein, by adjusting the distance between said transmission line elements, the interaction of the magnetic fields of said transmission line elements with an external sample can be controlled and optimized for nuclear magnetic resonance signal generation excitation.

37. The magnetic resonance imaging system of claim 34 wherein said series of transmission line elements decrease the inductance of each line element and minimize the electric fields associated therewith.

38. The magnetic resonance imaging system of claim 34 wherein said series of transmission line elements have inherent shielding.

39. The magnetic resonance imaging system of claim 34 wherein said transmit inductor system consists of N transmission line elements arranged in a geometric pattern in which each of said transmission line elements is substantially equidistant from each adjacent transmission line element.

40. The magnetic resonance imaging system of claim 39 wherein said geometric pattern is circular or elliptical.

41. The magnetic resonance imaging system of claim 39 wherein said geometric pattern is flat or curved.

42. The magnetic resonance imaging system of claim 34 wherein said outer and inner conductors include additional lumped or distributed capacitive or inductive circuit components.

43. The magnetic resonance imaging system of claim 34 wherein each of said transmission line elements couples to the other of said transmission line elements through mutual inductance and capacitive coupling.

44. The magnetic resonance imaging system of claim 34 wherein distributed impedance elements are connected between certain of said transmission line elements to alter the coupling therebetween.

45. The magnetic resonance imaging system of claim 34 wherein impedance elements are connected between said transmission line elements to establish interactions that establish frequency dependent relations between the currents and voltages present on certain of said transmission line elements.

46. The magnetic resonance imaging system of claim 34 wherein a given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

47. The magnetic resonance imaging system of claim 34 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use the generation of the NMR signal.

48. The magnetic resonance imaging system of claim 34 including RF power amplifiers coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal.

49. The magnetic resonance imaging system of claim 34 including at least an RF power amplifier reactively coupled to at least one of said transmission line elements for transferring energy into said coil during the generation of said transmit signal, the impedance of said RF power amplifier and the impedance of said one of said transmission line elements being matched.

50. The magnetic resonance imaging system of claim 34 wherein the phases of the current in a plurality of said transmission line elements are offset so as to create an elliptically polarized magnetic field for generating and/or detecting nuclear magnetic resonance signals.

51. The magnetic resonance imaging system of claim 34 including a plurality of diodes operatively connected to a plurality of said transmission line elements for tuning the coupling between said transmission line elements.

52. The magnetic resonance imaging system of claim 34 including coupling components between one or more of said transmission line elements to allow the currents on each of said transmission line elements to be independently controlled with separate power amplifiers connected to one or more of said transmission line elements allowing a transmit field of desired spatial intensity and phase to be generated.

53. The magnetic resonance imaging system of claim 34 with individual power amplifiers connected to each transmission line element with impedance mismatches designed to allow the current of each transmission line element to be independently controlled allowing a transmit field of desired spatial intensity and phase to be generated.

54. A magnetic resonance imaging system comprising: (a) a housing providing a medical diagnostic chamber for a subject therewithin lying along an axis; (b) a receive inductor system about said axis in proximity with said housing; (c) a gradient inductor system operatively associated with said receive inductor system; (d) a field inductor system operatively associated with said receive inductor system; (e) said receive inductor system constituting a coil having an outer surface about said axis and including a series of electrical transmission line elements paraxially distributed with respect to said axis about said subject, each of said transmission line elements including an outer conductor and an inner conductor, said inner conductor being spaced from said outer conductor in a direction perpendicular to said outer surface, said coil including a means for detuning said coil to prevent disturbance of the transmit fields generated by a separate transmit inductor system; (f) said coil receiving from said subject fields of magnetic resonance energy; (g) said gradient inductor system initiating perturbations in said fields and producing signals derived responsively from said perturbations; (h) said signals corresponding to spatial indicia derived from said subject.

55. The magnetic resonance imaging system of claim 54 wherein, by adjusting the distance between said transmission line elements, the interaction of the magnetic fields of said transmission line elements with an external sample can be controlled and optimized for nuclear magnetic resonance signal detection.

56. The magnetic resonance imaging system of claim 54 wherein said series of transmission line elements decrease the inductance of each transmission line element and minimize the electric fields associated therewith.

57. The magnetic resonance imaging system of claim 54 wherein said series of transmission line elements have inherent shielding.

58. The magnetic resonance imaging system of claim 50 wherein said series of transmission line elements are combined to optimize NMR signal reception.

59. The magnetic resonance imaging system of claim 50 wherein signals from said series of transmission line elements are combined to decode spatial information derived from the NMR signal.

60. The magnetic resonance imaging system of claim 50 wherein said receive inductor system consists of N transmission line elements arranged in a geometric pattern in which each of said transmission line elements is substantially equidistant from each adjacent transmission line element.

61. The magnetic resonance imaging system of claim 60 wherein said geometric pattern is circular or elliptical.

62. The magnetic resonance imaging system of claim 60 wherein said geometric pattern is flat or curved.

63. The magnetic resonance imaging system of claim 59 wherein said outer and inner conductors include additional lumped or distributed capacitive or inductive circuit components.

64. The magnetic resonance imaging system of claim 59 wherein each of said transmission line elements couples to the other of said transmission line elements through mutual inductance and capacitive coupling.

65. The magnetic resonance imaging system of claim 59 wherein distributed impedance elements are connected between certain of said transmission line elements alter the coupling therebetween.

66. The magnetic resonance imaging system of claim 59 wherein impedance elements are connected between said transmission line elements to establish interactions that establish frequency dependent relations between the currents and voltages present on certain of said transmission line elements.

67. The magnetic resonance imaging system of claim 59 wherein a=given current distribution is obtained on said transmission line elements at a given frequency by adjustment of the geometry of said transmission line elements and circuit components connected among said transmission line elements.

68. The magnetic resonance imaging system of claim 59 wherein the fields generated by the currents in said transmission line elements are superposed to create a given magnetic field configuration for use in the detection of the NMR signal.

69. The magnetic resonance imaging system of claim 59 including RF receivers coupled to at least one of said transmission line elements for transferring energy out of said coil during receive.

70. The magnetic resonance imaging system of claim 59 wherein the phases of the current in a plurality of said transmission line elements are offset so as to create an elliptically polarized magnetic field for detecting nuclear magnetic resonance signals.

71. The magnetic resonance imaging system of claim 69 including a plurality of diodes operatively connected to a plurality of said transmission line elements for tuning the coupling between said transmission line elements and said RF receivers.

72. The magnetic resonance imaging system of claim 59 including coupling elements between one or more of said transmission line elements in order to make the currents on each of said transmission line elements relatively independent allowing the signals from two or more of said transmission line elements to be optimally combined before or after image reconstruction.

73. The magnetic resonance imaging system of claim 59 with individual preamplifiers connected to each of said transmission line elements with impedance mismatches designed to allow each of said transmission line elements to operate independently allowing the signals from two or more of said transmission line element to be optimally combined either before or after image reconstruction.

74. The magnetic resonance imaging system of claim 59 with individual preamplifier/receivers connected to each transmission line element with the independent information obtained from individual transmission line elements being used to decode spatial information regarding said subject.

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1. Document ID: US 20030146750 A1

L14: Entry 1 of 26 File: PGPB Aug 7, 2003

PGPUB-DOCUMENT-NUMBER: 20030146750
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030146750 A1

TITLE: RF coil for imaging system

PUBLICATION-DATE: August 7, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Vaughan, J. Thomas JR.	Stillwater	MN	US	

US-CL-CURRENT: 324/318; 707/104.1

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

2. Document ID: US 20030038632 A1

L14: Entry 2 of 26 File: PGPB Feb 27, 2003

PGPUB-DOCUMENT-NUMBER: 20030038632
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030038632 A1

TITLE: Method and apparatus for enhanced multiple coil imaging

PUBLICATION-DATE: February 27, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Duensing, G. Randy	Gainesville	FL	US	
Varosi, Steve	Gainesville	FL	US	
King, Scott B.	Gainesville	FL	US	

US-CL-CURRENT: 324/307; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

3. Document ID: US 20030020476 A1

L14: Entry 3 of 26 File: PGPB Jan 30, 2003

PGPUB-DOCUMENT-NUMBER: 20030020476
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030020476 A1

TITLE: Method and apparatus for magnetic resonance imaging

PUBLICATION-DATE: January 30, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Duensing, G. Randy	Gainesville	FL	US	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

4. Document ID: US 20030020475 A1

L14: Entry 4 of 26 File: PGPB Jan 30, 2003

PGPUB-DOCUMENT-NUMBER: 20030020475
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20030020475 A1

TITLE: RF coil system for an MR apparatus

PUBLICATION-DATE: January 30, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Leussler, Christoph Guenther	Hamburg		DE	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

5. Document ID: US 20020125888 A1

L14: Entry 5 of 26 File: PGPB Sep 12, 2002

PGPUB-DOCUMENT-NUMBER: 20020125888
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020125888 A1

TITLE: Magnetic resonance imaging apparatus

PUBLICATION-DATE: September 12, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Visser, Frederik	Eindhoven		NL	
Haans, Paulus Cornelius Hendrikus Adrianus	Eindhoven		NL	
Van Den Brink, Johan Samuel	Eindhoven		NL	

US-CL-CURRENT: 324/318; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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6. Document ID: US 20020103429 A1

L14: Entry 6 of 26 File: PGPB Aug 1, 2002

PGPUB-DOCUMENT-NUMBER: 20020103429
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020103429 A1

TITLE: Methods for physiological monitoring, training, exercise and regulation

PUBLICATION-DATE: August 1, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
deCharms, R. Christopher	Moss Beach	CA	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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7. Document ID: US 20020087063 A1

L14: Entry 7 of 26 File: PGPB Jul 4, 2002

PGPUB-DOCUMENT-NUMBER: 20020087063
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020087063 A1

TITLE: New method and system for processing magnetic resonance signals to remove transient spike noise

PUBLICATION-DATE: July 4, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lou, Xiaoming	Waukesha	WI	US	

US-CL-CURRENT: 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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8. Document ID: US 20020042563 A1

L14: Entry 8 of 26 File: PGPB Apr 11, 2002

PGPUB-DOCUMENT-NUMBER: 20020042563
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020042563 A1

TITLE: Method and apparatus for objectively measuring pain, pain treatment and other related techniques

PUBLICATION-DATE: April 11, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Becerra, Lino R.	Cambridge	MA	US	
Breiter, Hans C.	Lincoln	MA	US	
Borsook, David	Concord	MA	US	

US-CL-CURRENT: 600/407; 600/411, 600/420, 600/427, 600/431, 600/473, 600/475

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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9. Document ID: US 6529000 B2

L14: Entry 9 of 26 File: USPT Mar 4, 2003

US-PAT-NO: 6529000
DOCUMENT-IDENTIFIER: US 6529000 B2

TITLE: Method and system for processing magnetic resonance signals to remove transient spike noise

DATE-ISSUED: March 4, 2003

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lou; Xiaoming	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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10. Document ID: US 6501274 B1

L14: Entry 10 of 26 File: USPT Dec 31, 2002

US-PAT-NO: 6501274
DOCUMENT-IDENTIFIER: US 6501274 B1

TITLE: Magnetic resonance imaging system using coils having paraxially distributed transmission line elements with outer and inner conductors

DATE-ISSUED: December 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ledden; Patrick	Malden	MA		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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11. Document ID: US 6452390 B1

L14: Entry 11 of 26 File: USPT Sep 17, 2002

US-PAT-NO: 6452390
DOCUMENT-IDENTIFIER: US 6452390 B1

TITLE: Magnetic resonance analyzing flow meter and flow measuring method

DATE-ISSUED: September 17, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wollin; Ernest	Leesburg	FL		

US-CL-CURRENT: 324/306; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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12. Document ID: US 6289232 B1

L14: Entry 12 of 26 File: USPT Sep 11, 2001

US-PAT-NO: 6289232
DOCUMENT-IDENTIFIER: US 6289232 B1

TITLE: Coil array autocalibration MR imaging

DATE-ISSUED: September 11, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jakob; Peter M.	Brookline Village	MA		
Sodickson; Daniel K.	Cambridge	MA		
Griswold; Mark	Brookline	MA		

US-CL-CURRENT: 600/410; 324/307, 324/309, 324/318, 324/322, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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13. Document ID: US 5977770 A

L14: Entry 13 of 26 File: USPT Nov 2, 1999

US-PAT-NO: 5977770
DOCUMENT-IDENTIFIER: US 5977770 A

TITLE: MR imaging of synchronous spin motion and strain waves

DATE-ISSUED: November 2, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ehman; Richard L.	Rochester	MN		

US-CL-CURRENT: 324/318; 600/421

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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14. Document ID: US 5910728 A

L14: Entry 14 of 26 File: USPT Jun 8, 1999

US-PAT-NO: 5910728
DOCUMENT-IDENTIFIER: US 5910728 A

TITLE: Simultaneous acquisition of spatial harmonics (SMASH): ultra-fast imaging with radiofrequency coil arrays

DATE-ISSUED: June 8, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel Kevin	Cambridge	MA		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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15. Document ID: US 5898306 A

L14: Entry 15 of 26 File: USPT Apr 27, 1999

US-PAT-NO: 5898306
DOCUMENT-IDENTIFIER: US 5898306 A
**** See image for Certificate of Correction ****

TITLE: Single circuit ladder resonator quadrature surface RF coil

DATE-ISSUED: April 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Haiying	Minneapolis	MN		
Truwit; Charles L.	Wayzata	MN		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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16. Document ID: US 5757187 A

L14: Entry 16 of 26 File: USPT May 26, 1998

US-PAT-NO: 5757187
DOCUMENT-IDENTIFIER: US 5757187 A

TITLE: Apparatus and method for image formation in magnetic resonance utilizing weak time-varying gradient fields

DATE-ISSUED: May 26, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wollin; Ernest	Leesburg	FL		

US-CL-CURRENT: 324/306; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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17. Document ID: US 5592085 A

L14: Entry 17 of 26 File: USPT Jan 7, 1997

US-PAT-NO: 5592085
DOCUMENT-IDENTIFIER: US 5592085 A
**** See image for Certificate of Correction ****

TITLE: MR imaging of synchronous spin motion and strain waves

DATE-ISSUED: January 7, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ehman; Richard L.	Rochester	MN		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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18. Document ID: US 5560360 A

L14: Entry 18 of 26 File: USPT Oct 1, 1996

US-PAT-NO: 5560360
DOCUMENT-IDENTIFIER: US 5560360 A
**** See image for Certificate of Correction ****

TITLE: Image neurography and diffusion anisotropy imaging

DATE-ISSUED: October 1, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Filler; Aaron G.	Seattle	WA		
Tsurda; Jay S.	Mercer Island	WA		
Richards; Todd L.	Seattle	WA		
Howe; Franklyn A.	London			GB2

US-CL-CURRENT: 600/408; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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19. Document ID: US 5539312 A

L14: Entry 19 of 26 File: USPT Jul 23, 1996

US-PAT-NO: 5539312
DOCUMENT-IDENTIFIER: US 5539312 A

TITLE: Detection and measurement of motion during NMR imaging using orbital navigator echo signals

DATE-ISSUED: July 23, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Fu; Zhuo F.	New York	NY		
Ehman; Richard L.	Rochester	MN		

US-CL-CURRENT: 324/309; 600/410

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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20. Document ID: US 5525906 A

L14: Entry 20 of 26 File: USPT Jun 11, 1996

US-PAT-NO: 5525906
DOCUMENT-IDENTIFIER: US 5525906 A

TITLE: Detection and elimination of wide bandwidth noise in MRI signals

DATE-ISSUED: June 11, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Crawford; Carl R.	Milwaukee	WI		
Eash; Matthew G.	Oconomowoc	WI		
Souza; Steven P.	Williamstown	MA		
Pelc; Norbert J.	Los Altos	CA		
DallaPiazza; Dennis G.	Pewaukee	WI		
Small; Daniel S.	Hartland	WI		
Stormont; Robert S.	Waukesha	WI		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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21. Document ID: US 5284144 A

L14: Entry 21 of 26 File: USPT Feb 8, 1994

US-PAT-NO: 5284144
DOCUMENT-IDENTIFIER: US 5284144 A

TITLE: Apparatus for hyperthermia treatment of cancer

DATE-ISSUED: February 8, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Delannoy; Jose	Monsen Baroeul			FR
Le Bihan; Denis	Rockville	MD		
Chen; Ching-nien	Catonsville	MD		
Levin; Ronald L.	Olney	MD		
Turner; Robert	Bethesda	MD		

US-CL-CURRENT: 600/412; 324/315, 600/422, 607/154

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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22. Document ID: US 5212450 A

L14: Entry 22 of 26 File: USPT May 18, 1993

US-PAT-NO: 5212450
DOCUMENT-IDENTIFIER: US 5212450 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: May 18, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hills	PA		
Srinivasan; Ravi	Philadelphia	PA		
Carvajal; Lucas	North Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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23. Document ID: US 5202635 A

L14: Entry 23 of 26 File: USPT Apr 13, 1993

US-PAT-NO: 5202635
DOCUMENT-IDENTIFIER: US 5202635 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: April 13, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Philadelphia	PA		
Murphy-Boesch; Joseph	Lafayette Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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24. Document ID: US 5194811 A

L14: Entry 24 of 26 File: USPT Mar 16, 1993

US-PAT-NO: 5194811
DOCUMENT-IDENTIFIER: US 5194811 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: March 16, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hill	PA		
Srinivasan; Ravi	Philadelphia	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 25. Document ID: US 5184076 A

L14: Entry 25 of 26 File: USPT Feb 2, 1993

US-PAT-NO: 5184076
DOCUMENT-IDENTIFIER: US 5184076 A

TITLE: Magnetic resonance imaging coil for applying NMR and ESR pulses

DATE-ISSUED: February 2, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ehnholm; Gosta J.	Helsinki			FI

US-CL-CURRENT: 324/318; 324/316

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 26. Document ID: US 5050605 A

L14: Entry 26 of 26 File: USPT Sep 24, 1991

US-PAT-NO: 5050605
DOCUMENT-IDENTIFIER: US 5050605 A
**** See image for Certificate of Correction ****

TITLE: Magnetic resonance imaging antennas with spiral coils and imaging methods employing the same

DATE-ISSUED: September 24, 1991

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Eydelman; Gregory	West Hempstead	NY		
Giambalvo; Anthony	Kings Park	NY		
Damadian; Raymond V.	Woodbury	NY		

US-CL-CURRENT: 600/422; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
ARRAY	558960
ARRAYS	154287
PROXIMITY	287650
PROXIMITIES	591
PROXIMITYS	0
OTHER	47
OTHERS	533404
NEIGHBOR$6	0
NEIGHBOR	16045
NEIGHBORARRAY	2
NEIGHBORBATCH	1
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3	US 6556013	☐
4	US 6515479	☐
5	US 6504369	☐

	R trieval Classif	Inventor	S	C	P	2	3	4	5
1		Felmlee, Joel P. et al.	☒	☐	☐	☐	☐	☐	☐
2		Yeung, David Ka Wai	☒	☐	☐	☐	☐	☐	☐
3		Withers, Richard S.	☒	☐	☐	☐	☐	☐	☐
4		Arz, Winfried et al.	☒	☐	☐	☐	☐	☐	☐
5		Varjo, Tomi E. K. et al.	☒	☐	☐	☐	☐	☐	☐

	Title	Current OR	Current XRef
1	Autocorrection of MR images acquired using phased-array coils	324/309	324/307; 324/312
2	Birdcage resonator for MRI system	324/318	324/322
3	Planar NMR coils with localized field-generating and capacitive elements	324/322	324/318
4	Switchable longitudinal gradient coil	324/318	324/322
5	Decoupling two or more channels on RF coil systems	324/318	324/322

	U	1	Document ID	Issue Date	Pages
1	☐	☒	US 6469506 B1	20021022	8
2	☐	☒	US 6100691 A	20000808	7
3	☐	☒	US 6556013 B2	20030429	17
4	☐	☒	US 6515479 B1	20030204	9
5	☐	☒	US 6504369 B1	20030107	8